Proposal of Gait Monitoring Shoes Based on Comparative Analysis of Walking Gait Cycle between Normal People and Stroke Patients

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***Abstract*— Elderly people have an increased risk of falling and consequently injuring themselves. The prevention of falls is one of major important issues to keep healthy life, because a fall is a main reason for them to hospital administration or a residential care. Motion capture systems are a key component to prevent falls. We comparably analyzed walking gait cycle between normal people and stroke patients using a wearable device (WD) to detect warning signs of falls. In this paper, we experimentally clarify to which body part a WD is best attached and what kind of signs are useful to prevent falls. We also develop a pair of shoes with a WD in each heel in accordance with results of pre-experiment.**

Keywords-falls; trip; wearable device; hemiplegia; guit; shoes.

I. INTRODUCTION

Percentage of elderly people in the population is increasing in the world [1]. As the number of elderly increases, the number of functionally impaired people, such as cerebrovascular patients who have one side paralysis, will also increase. These people have an increased risk of falling and consequently injuring themselves [2]. Falling down is one if the main reasons for them to be hospitalized or placed in residential care.

There are many researches related to a fall for elder people. The World Health Organization Regional Office for Europe analyzed these studies and classified fall risks amongst elderly people by history of falls, age, gender, living alone, ethnicity, medicines, medical conditions, impaired mobility and gait, sedentary behavior, psychological status - fear of falling, nutritional deficiencies, visual impairments, and foot problems [2]. Stroke patients, such as a cerebrovascular disease especially are at substantially high risk of falling [3][4][5][6]. The higher frequency of falls for them due to weak muscles, one-side paralysis, and toes pointing down. For people with impaired mobility and gait, tripping is a major cause of falls [7][8], so we focus on tripping in this paper.

Since weak muscles, one-side paralysis, and toes pointing down strongly appear in the movement of leg and foot, motion capture for them are a key component to analyze impaired mobility and gait and useful to prevent tripping and conduct therapy and rehabilitation of hemiplegia.

Weijun Tao et al. reviewed gait analysis technologies based on wearable sensors that were the accelerometer, gyroscope, electromagnetic tracking system, magneto-resistive sensors, flexible goniometer, sensing fabric, force sensor, and sensors for electromyography. They mentioned that fall risk estimation is an important application of gait analysis using wearable sensors [9]. However, they did not describe about motion of gait for elderly people or stroke patients.

Stacy J. Morris Bamberg et al. developed a prototype shoe in which several kinds of wearable sensors, such as accelerometer, gyroscope, force sensor, bidirectional bend sensor and so on [10]. The calibrated sensor outputs were almost same as results obtained simultaneously from a biological motion measuring equipment. They calculated the maximum pitch (angle between the shoe sole and floor at the toe-off timing), minimum pitch (angle between the shoe sole and floor at the heel-strike timing), the stride length from output of accelerometers and gyroscopes. They also compared the maximum pitch, minimum pitch and stride length between the healthy gait and parkinsonian gait. There were differences on mean value of calculated data between the healthy gait and parkinsonian gait. However, considering standard deviation of calculated data, such differences were small. They also did not measure and analyze motions of gait for elderly people or stroke patients.

Farzin Dadashi et al. measured motion of gait for many elderly people with shoe-worn inertial sensors and provided normative values for a clinician to measure reference gait parameters [11]. They analyzed motion of gait and clarified the difference in gait parameters, such as the clearance between a shoe sole and floor, gait speed, stride length between males and females by considering the effect of age factors. However, their data did not show differences clearly between the male and female, and the effect of age factor. And, they did not investigate data for stroke patients or analyze reasons for tripping.

Mourad Benoussaad et al. introduced a method to robustly estimate foot clearance during walking using a single inertial measurement unit (IMU) placed on the subject's foot [12]. In their paper, the foot clearance was the height of ankle from a floor. However, the toe clearance is more critical for tripping. And, they did not measure the toe clearance for stroke patient or analyze reasons for tripping.

Here, we focus on extracting warning signs of tripping for stroke patients, such as cerebrovascular patients. We got output data of an acceleration sensor and gyroscope sensor in the wearable device (WD), Sony SmartWatch 3, mounted on the front part of a foot to estimate the kicking power and change of angle between a foot and floor. We noticed that an

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angle velocity at the terminal stance and an angle at the terminal swing show clearly differences between unimpaired student and stroke patients. After that, we develop a pair of shoes built a WD on.

In the next section, we consider how people trip on a flat floor. Different features between physically unimpaired students and stroke patients are extracted from measured data in Section III. The best body part to which a wearable device (WD) can be attached is described in Section IV. We introduce a pair of shoes with a WD in each heel and example data measured using them in Section V. Conclusions are summarized in Section VI.

II. CONSIDERATION OF TRIPPING FACTOR

When the swing foot progression is unexpectedly obstructed, a trip occurs that leads to a forward rotation of the body and eventually might cause a fall.

Mourad Benoussaad et al. measured the minimum toe clearance (MinTC) to avoid tripping. MinTC is a critical value to clear obstacles on the ground or floor. However, elderly people, especially those who have had strokes, sometimes trip on flat ground or floors, not obstacles. In this section, we consider reasons a person trips on flat ground or floors. We divide the normal walking gait cycle into eight phases the same as Weijun Tao et al. as shown in Fig. 1 [9]: (1) initial contact (heel-strike timing), (2) loading response, (3) mid-stance, (4) terminal stance (toe-off timing), (5) pre-swing, (6) initial swing, (7) mid-swing, and (8) terminal swing.

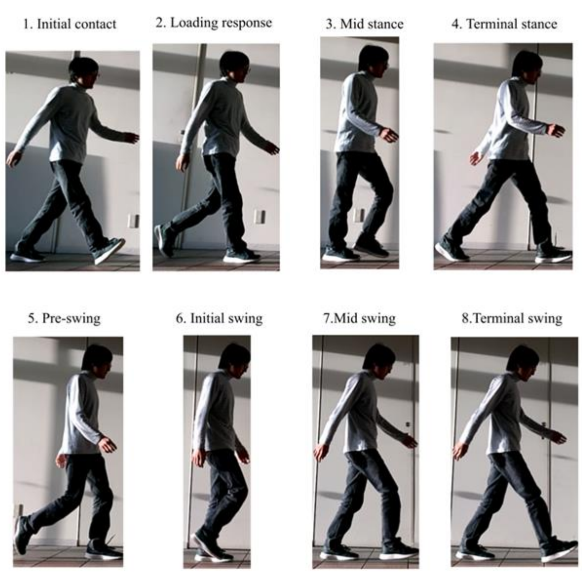


Figure 1. Normal walking gait cycle (See a right foot)

Most stroke patients have weak muscles and are hard to raise their toe. Therefore, they are at risk of three types of trips.

- Case 1: A toe touches the floor first instead of a heel at phase 1. Since phases 2-5 are skipped, the toe is dragged along the floor. When the dragging strength is stronger than the person's muscular power, he/she trips (Fig. 2(a)).
- Case 2: Kicking power of the front part of a foot is insufficient at phases 4 and 5 to raise the heel and toe up from the floor. In this case, a person does not swing but shuffles. When the frictional force between a shoe sole and the ground or floor is stronger than his/her muscular power, he/she trips (Fig. 2(b)).
- Case 3: A toe touches the floor due to it pointing down during the swing phases (5-8), and the knee goes further forward than the foot. When the dragging strength is stronger than the person's muscular power, he/she trips (Fig. 2(c)).

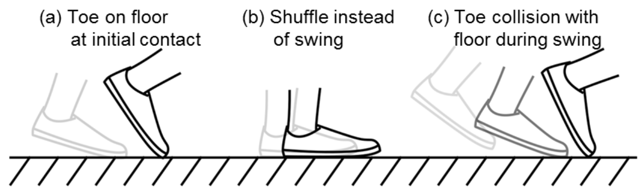


Figure 2. Cases of tripping

The above problems suggest that the kicking power at phases 4 and 5 and the angle between the foot and a floor are critical parameters.

III. EXTRACTION OF WARNING SIGNS FOR TRIPPING

In this section, we experimentally investigate whether the kicking power at phases 4 and 5, and the angle between the foot and lower limb are critical parameters.

A. Experimental method

Since kicking power must be expressed as the angle velocity or the acceleration for the foot, we mounted a WD which had an accelerometer and gyroscope on the foot. In this experiment, we used Sony SmartWatch 3 as a WD which is mounted on the front part of a foot with Velcro tape as shown in Fig. 3. This mounting position was same as one in Farzin Dadashi's experiment [11]. The sampling rate was 40 msec.

We measure angle velocity for up and down directions of the front part of the foot (X axis of a 3D gyroscope). We also adopted a three-point moving average of the angle velocity to calculate the angle, because output values extremely change up and down. Therefore, the angle for X axis $Angle_{xn}$ at time t_n is calculated as follows.

$$Angle_{xn} = Angle_{xn-1} + \frac{t_n - t_{n-1}}{1000} \times \frac{G_{n-1} + G_n + G_{n+1}}{3} \quad (1)$$

G_{xn} is the value of angle velocity for X axis at time t_n. We also recorded video and measured locations of a patient's knee, ankle and foot by using Microsoft-KINECT to monitor their motions (Fig. 4). The UNIX time is introduced to synchronously measure data with a WD and MS-KINECT.

Before measuring, we investigated the measuring accuracy of Sony SmartWatch 3 using a slant rule as shown in Fig. 5. We measured data five times. Calculated angles vs. angles given by the slant rule are listed in Table I. These data showed calculated angles were so accurate. We noticed drift errors of a gyroscope that increase the value by 0.2 rad. /sec. during a WD sets on a flat floor. However, each measurement lasted less than 20 sec. Therefore, we think the effect of the drift error is negligible.

Participants were five physically unimpaired students and five stroke patients. Every stroke patients in this experiment had one-side paralysis, and trained periodically at a rehabilitation facility. Some of them used a wheel chair and could not walk by himself before training. They walked along a straight line to MS-KINECT. A WD was attached on the front part of foot on the paralysis side. We measured data for each stroke patient two times.

(a) WD: Sony SmartWatch 3 (b) WD mounted on foot

Figure 3. Measuring device and WD mounting method

(a) MS-KINECT set up in rehabilitation facility (b) WD mounted on stroke patient

Figure 4. Measurement environment in rehabilitation facility

Figure 5. Slant rule

TABLE I. ACCURACY OF CALCULATED ANGLES

Given angle (degree)	Calculated angle (degree)	SD (degree)
+50	+49.00	0.45
+40	+39.17	0.74
+30	+28.92	0.51
+20	+19.58	0.63
+10	+9.21	0.58
0	0.21	0.15
-10	-9.72	0.58
-20	-19.47	0.34
-30	-30.70	0.56
-40	-40.57	0.41
-50	-50.54	0.53

B. Measured data and consideration

Fig. 6 and 7 show examples of change of angle velocity, angle, and acceleration for a physically unimpaired student and a stroke patient. Data for two steps are plotted.

Each flat period (roughly the center period) in these figures is when the entire shoe sole touched the floor; this period corresponds to phases 2 (loading response) and 3 (mid-stance). The reason that the value during this period is not zero is that the WD measures the angle between the front part of the foot and the floor, which depends on the person and shoe. Therefore, we reset this angle for the gait monitoring shoes described in Sections IV and V when the entire shoe sole touched the floor. This processing enables the WD to measure the angle between the back of the foot and the floor. This value does not depend on person or shoe.

The maximum angle velocity at timing A means the kicking power from phase 4 (terminal stance) to 5 (pre-swing), and the minimum angle at timing B means the angle to the floor at phase 8 (terminal swing). This processing removes the drift error of the gyroscope.

Lower angle velocity at A in Fig. 6 is about 420 deg./sec. On the other hand, higher angle velocity at A in Fig. 7 is about 250 deg./sec. Thus, a physically unimpaired student and a stroke patient obviously differ in terms of gait. The stroke patient clearly has weaker kicking power at phase 4 (terminal stance) than the physically unimpaired student.

Higher angle at B in Fig. 6 is about -18 degree. On the other hand, lower angle at B in Fig. 7 is about -8 degree. Thus, a physically unimpaired student and a stroke patient obviously differ in terms of the angle to a floor at phase 8 (terminal swing). This shows that it is difficult for a stroke patient to raise his or her toe at the terminal swing phase.

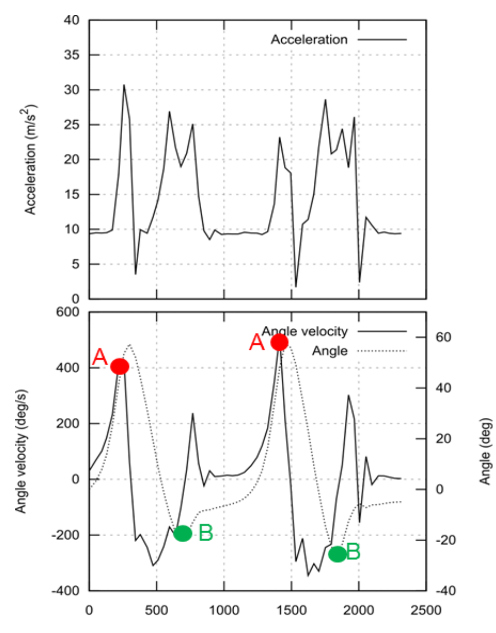


Figure 6. Changes of angle velocity, angle, and acceleration for physically unimpaired student

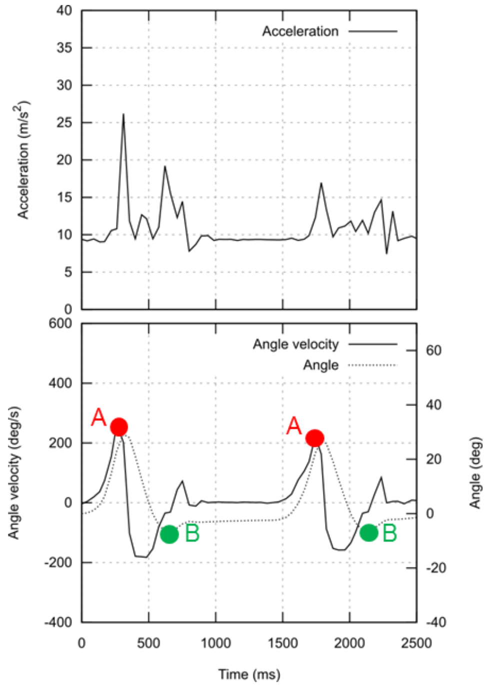


Figure 7. Changes of angle velocity, angle, and acceleration for stroke patient

The other hand, the acceleration basically change corresponding to the angle velocity and angle. However, they have much noise, and their amplitudes are not stable.

Tables II and III list the averages and standard deviations (SDs) of measured data for angle velocity at timing A and angle at timing B. The angle velocity at timing A is clearly different between unimpaired students and stroke patients. There is a big difference between them in the angle at timing B, however this value would be sometimes overlapped each other.

We also measure the cadence for a gait using the WD and MS-KINECT. In this paper, we define the cadence as the number of steps per minute. We estimated the cadence derived from an average of 10 intervals between one timing A and the next A, which are peak angle velocities of a step, when a WD is used. Estimated cadence is listed in Table IV. A physically unimpaired student and stroke patient do not obviously differ in terms of cadence.

As the result of this experiment, we decided to adopt the change of angle velocity and angle between a foot and floor to detect warning signs of falls.

TABLE II. ANGLE VELOCITY AT THE TERMINAL STANCE

Participant	Average (deg./s)	SD (deg./s)
Student	509.36	18.91
Stroke patient	342.06	86.52

Table III. Angle at the terminal swing

Participant	Average (deg.)	SD (deg.)
Student	-17.76	8.02
Stroke patient	-7.45	8.02

TABLE IV. ESTIMATED CADENCE

Participant	Average (steps/m)	SD (steps/m)
Student	46.4	5.0
Stroke patient	49.0	6.2

IV. SUITABLE WD ATTACHING POSITION

A WD has to be attached somewhere on a body during walking to detect signs of tripping to prevent a fall. A WD was attached to the front part of the foot in Section III. However, it is difficult for a WD to firmly be set at this place for a long time because it is easily detached. Therefore, we studied which position is the best to detect the change of angle velocity for a foot and angle between a foot and floor. We attached WDs to a heel and a lower limb as shown in Fig. 8.

For this test, we used STEVAL-WESU1 by STMicro-electronics (See Fig. 9) as a WD instead of Sony SmartWatch 3. This wearable unit includes four sensors:

- 3D-accelerometer,
- 3D-gyroscope,
- 3D-magnetometer,
- MEMS pressure.

This device is 37 x 40 x 8 mm and weighs 9.6 g.

We inserted STEVAL-WESU1 into the heel of a shoe as shown in Fig. 8 (a) (details in Section V). Angle velocity and acceleration data of STEVAL-WESU1 are sent to and processed by an Android smartphone. The sampling rate was 40 ms. We adopt a three-point moving average to remove noise.

We requested three unimpaired students to walk with their normal gait. Since their data change was basically the same, graphs of one participant are shown in Fig. 10 and 11. Both plotted lines in Fig. 10 are similar in shape to those in Fig. 6. Timing A and B correspond to timing A and B in Fig. 6. Timing B in Fig. 10, which is the angle at the terminal swing, is shown more clearly than that in Fig. 6. On the other hand, timing C in Fig. 11 shows the kicking power from phase 4 (terminal stance) to 5 (pre-swing) is the same as timing A in Fig. 6 and 10. However, the angle at timing D in Figure 11 is between not the foot and floor but a single limb and the vertical line to the floor. The plotted angle in Fig. 11 clearly shows a change of angle for the single limb.

As the result of this experiment, we decided that the heel was the best position to place a WD.

(a) Heel (b) Single limb

Figure 8. WD attaching position

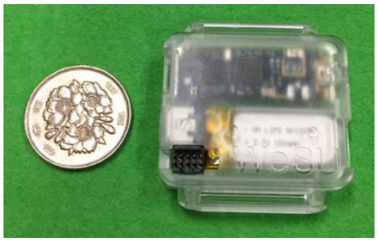


Figure 9. STEVAL-WESU1 by STMicroelectronics

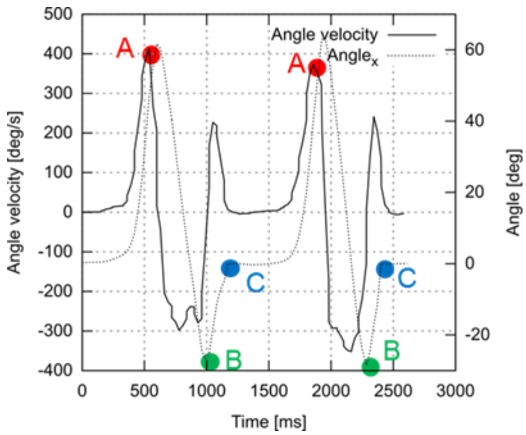


Figure 10. Angle velocity and angle data at heel in normal walk

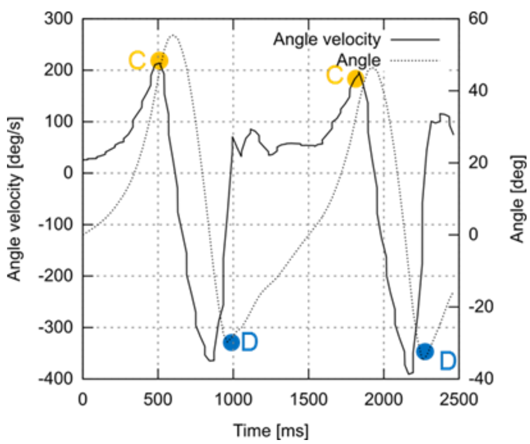


Figure 11. Angle velocity and angle data at single limb in normal walk

V. GAIT MONITORING SHOES

As described in Section IV, we determined the heel of a shoe is the best place to measure angle velocity of the foot and angle between the back part of a foot and the floor. We inserted a WD (STEVAL-WESU1 by STMicroelectronics) into soles of both shoes. And, we also developed a gait monitoring application for Android smartphone which measures and stores the angle velocity and angle as shown in Fig. 12. The upper part shows ID of WD for the right and left shoe, and the lower part shows angle velocities at A in each step for right foot, angle at B in each step for right foot, angle velocities at A in each step for left foot, and angle at B in each step for left foot. In this application, direction of angle is turned. When these graphs were measured, a participant played a stroke patient who had a one-side paralysis for the right side of the body. Therefore, most strength of angle velocity at A for right foot were smaller than that for left foot. And, most amplitude of angle at B for right foot were smaller than that for left foot.

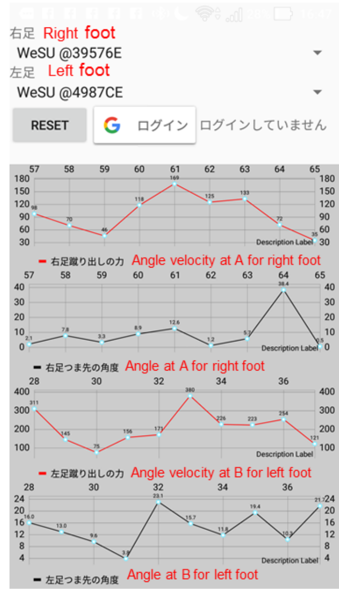
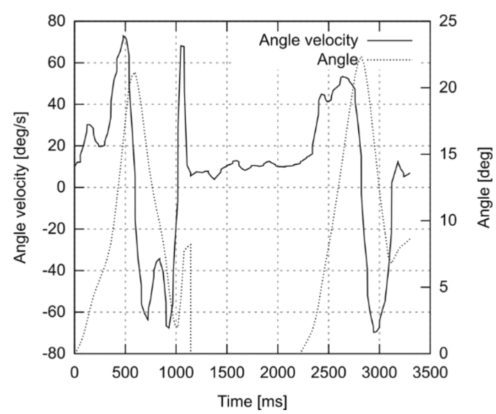


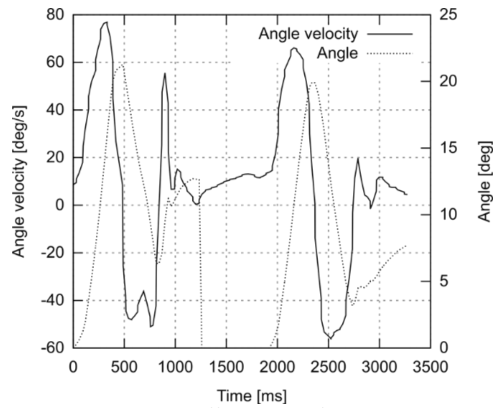
Figure 12. Gait monitoring application for Android

We experimentally monitored walking for two participants. They were unimpaired people. They walked and played the three types of trips in Fig. 2. Example measured data are shown in Fig. 13. Therefore, curves of angles in these graphs have different discontinuity to those in other graphs at the sole of a shoe touching a floor.

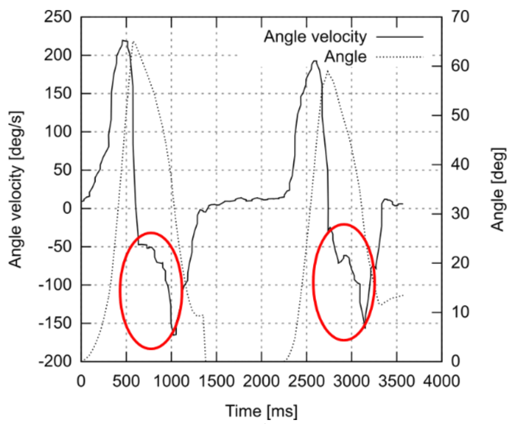
In (1); toe touching the floor first instead of a heel, and (2); shuffling, shapes of angle velocity resemble that of the normal walk shown in Fig. 10. However, maximum values of angle velocity and minimum angle in a cycle in Fig. 13 (1) and (2) are much smaller than those in Fig. 10. Their absolute minimum values are also much smaller than those in Fig. 10. This feature must show that when muscle strength is weaker, more trips occur.



(1) Toe on floor at the initial contact



(2) Shuffle instead of swing



(3) Toe collision with floor at the initial swing

Figure 13. Example measured data for tripping with gait monitoring shoes

In (3), a toe dragging the floor due to it pointing down at the initial swing, the swing did not start smoothly. The red circle in Fig. 13 (3) shows this situation clearly. In the case of a normal walk, angle velocity rapidly decrease from the pre-swing to the initial swing. However, in (3), the angle velocity limply decrease on the way.

VI. CONCLUSION

We comparably analyzed walking gait cycle between normal people and stroke patients using a wearable device (WD) to detect warning signs of falls. The results of experiments for physically unimpaired students and stroke patients show that the angle velocity at the terminal stance phase and the angle between the back of foot and a floor at the terminal swing are critical parameters. We determined a shoe heel to be the best place to place a WD, inserted WDs into heels of a pair of shoes, and showed data measured using them. We plan to develop a warning system using the proposed shoes for fall prevention. More data of above critical parameters for stroke patients in daily life including tripping conditions will be measured to develop such system.

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